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For and on behalf of RWS Group Ltd

The 19th day of April 2005



Description

Method for producing and applying an antiscatter grid
5 or collimator to an x-ray or gamma detector

The present invention relates to a method for producing
and applying an antiscatter grid or collimator to an x-
ray or gamma detector having matricially arranged
10 detector elements which form a detector surface with
detection regions sensitive to x-radiation and/or gamma
radiation and less sensitive intermediate regions and
to an x-ray and gamma detector having an antiscatter
grid or collimator which has been produced and applied
15 using this method.

In radiography, stringent requirements are currently
placed on the image quality of the x-ray images. In
such images, as are taken especially in medical x-ray
20 diagnosis, an object to be studied is exposed to x-
radiation from an approximately point radiation source,
and the attenuation distribution of the x-radiation is
registered two-dimensionally on the opposite side of
the object from the x-ray source. Line-by-line
25 acquisition of the x-radiation attenuated by the object
can also be carried out, for example in computer
tomography systems. Besides x-ray films and gas
detectors, solid-state detectors are being used
increasingly as x-ray detectors, these generally having
30 a matricial arrangement of optoelectronic semiconductor
components as photoelectric receivers. Each pixel of
the x-ray image should ideally correspond to the
attenuation of the x-radiation by the object on a
straight axis from the point x-ray source to the
35 position on the detector surface corresponding to the
pixel. X-rays which strike the x-ray detector from the
point x-ray source in a straight line on this axis are
referred to as primary beams.

The x-radiation emitted by the x-ray source, however, is scattered in the object owing to inevitable interactions, so that, in addition to the primary beams, the detector also receives scattered beams, so-called secondary beams. These scattered beams, which, depending on the properties of the object, can cause up to 90% or more of the total signal response of an x-ray detector in diagnostic images, constitute an additional noise source and therefore reduce the identifiability of fine contrast differences. This substantial disadvantage of scattered radiation is due to the fact that, owing to the quantum nature of the scattered radiation, a significant additional noise component is induced in the image recording.

15 In order to reduce the scattered radiation components striking the detectors, so-called antiscatter grids are therefore interposed between the object and the detector. Antiscatter grids consist of regularly arranged structures that absorb the x-radiation, between which transmission channels or transmission slits for minimally attenuated transmission of the primary radiation are formed. These transmission channels or transmission slits, in the case of focused antiscatter grids, are aligned with the focus of the x-ray tube according to the distance from the point x-ray source, that is to say the distance from the focus. In the case of unfocused antiscatter grids, the transmission channels or transmission slits are oriented perpendicularly to the surface of the antiscatter grid over its entire area. However, this leads to a significant loss of primary radiation at the edges of the image recording, since a sizeable part of the incident primary radiation strikes the absorbing regions of the antiscatter grid at these points.

In order to achieve a high image quality, very stringent requirements are placed on the properties of

x-ray antiscatter grids. The scattered beams should, on the one hand, be absorbed as well as possible, while on the other hand, the highest possible proportion of primary radiation should be transmitted unattenuated through the antiscatter grid. It is possible to achieve a reduction of the scattered beam component striking the detector surface by a large ratio of the height of the antiscatter grid to the thickness or diameter of the transmission channels or transmission slits, that is to say by a high aspect ratio. The thickness of the absorbing structure elements or wall elements lying between the transmission channels or transmission slits, however, can lead to image perturbations by absorption of part of the primary radiation. Specifically when solid-state detectors are used, inhomogeneities of the grids, that is to say deviations of the absorbing regions from their ideal position, cause image perturbations by projection of the grids in the x-ray image.

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In order to minimize image perturbations due to antiscatter grids, it is known to move the grids in a lateral direction during the recording. In the case of very short exposure times of, for example, 1 - 3 ms, however, stripes may also occur in the image if the movement speed of the grids is insufficient. Even in the event of very long exposure times, perturbing stripes may occur owing to reversal of the grid movement direction during exposure.

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In recording x-ray images, increasing use has recently been made of solid-state detectors which are formed from a plurality of matricially arranged detector elements. The detector elements are arranged in this case in a generally square or rectangular grating. In the case of such solid-state detectors, as well, there is a need to employ effective suppression measures to reduce as far as possible the striking of scattered

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beams on the detector surface formed by the detector elements. Because of the regular structuring of the pixels, formed by the detector elements, of the detector, there is here, in addition, the risk of mutual interference between the structures of pixels and antiscatter grids. Disturbing moiré phenomena can thereby arise. These can certainly in specific instances be minimized or removed by a downstream image processing measure. However, this is possible only when their projection image on the detector is absolutely immutable.

The same problem occurs in nuclear medicine, especially when using gamma cameras, for example Anger cameras. With this recording technique also, as with x-ray diagnosis, it is necessary to ensure that the fewest possible scattered gamma quanta reach the detector. In contrast to x-ray diagnosis, the radiation source for the gamma quanta lies inside the object in the case of nuclear diagnosis. In this case, the patient is injected with a metabolic preparation labeled with particular unstable nuclides, which then becomes concentrated in a manner specific to the organ. By detecting the decay quanta correspondingly emitted from the body, a picture of the organ is then obtained. The profile of the activity in the organ as a function of time permits conclusions about its function. In order to obtain an image of the body interior, a collimator that sets the projection direction of the image needs to be placed in front of the gamma detector. In terms of functionality and structure, such a collimator corresponds to the antiscatter grid in x-ray diagnosis. Only the gamma quanta dictated by the preferential direction of the collimator can pass through the collimator, and quanta incident obliquely to it are absorbed in the collimator walls. Because of the higher energy of gamma quanta compared with x-ray quanta,

collimators need to be made many times higher than antiscatter grids for x-radiation.

For instance, scattered quanta may be deselected during the image recording by taking only quanta with a particular energy into account in the image. However, each detected scattered quantum entails a dead time in the gamma camera of, for example, one microsecond, during which no further events can be registered. Therefore, if a primary quantum arrives shortly after a scattered quantum has been registered, it cannot be registered and it is lost from the image. Even if a scattered quantum coincides temporally - within certain limits - with a primary quantum, a similar effect arises. Since the evaluation electronics can then no longer separate the two events, too high an energy will be determined and the event will not be registered. Both said situations explain how highly effective scattered beam suppression leads to improved quantum efficiency in nuclear diagnosis as well. As the end result, an improved image quality is thereby achieved for equal dosing of the applied radionuclide or, for equal image quality, a lower radionuclide dose is made possible, so that the patient's beam exposure can be reduced and shorter image recording times can be achieved.

In future, increasing use will also be made for recording gamma images of solid-state detectors which are formed from a plurality of matrixially arranged detector elements. The detector elements are arranged in this case in a generally square or rectangular grating. In the case of such solid-state detectors, as well, there is a need to employ effective suppression measures to reduce as far as possible the striking of scattered beams on the detector surface formed by the detector elements. Because of the regular structuring of the pixels, formed by the detector elements, of the

detector, there is here, in addition, the risk of mutual interference between the structures of pixels and collimators.

5 Collimators for gamma cameras are generally produced from mechanically folded lead lamellae. This is a relatively cost-efficient solution, although it has the disadvantage that, in particular when using solid-state cameras with matricially arranged detector elements,
10 for example in the case of cadmium-zinc telluride detectors, perturbing aliasing effects can arise because the structure of these collimators is then relatively coarse.

15 The publication by G. A. Kastis et al., "A Small-Animal Gamma-Ray Imager Using a CdZnTe Pixel Array and a High Resolution Parallel Hole Collimator" discloses a method for producing a cellularly constructed collimator for gamma radiation. In this case, the collimator is
20 produced from laminated layers of metal films, here made of tungsten, which are photochemically etched. However, on account of the large number of photolithographic exposure and etching steps, this production method is very elaborate and cost-intensive.

25 US 6,021,173 A describes an approach which is intended to avoid moiré structures during operation of an x-ray detector having matricially arranged detector elements in conjunction with an antiscatter grid arranged in a
30 stationary fashion. In this publication, the antiscatter grid is applied directly to the x-ray detector over the detector surface. The absorbing structure elements of the antiscatter grid are designed at a spacing from one another which is smaller than the
35 extent of the smallest resolvable detail in the x-ray image. The regularly arranged absorbing structure elements are consequently formed at so high a spatial frequency that they are situated beyond the resolving

power of the x-ray detector. Since the spacing of the structure elements in the antiscatter grid cannot be selected to be arbitrarily small, it is necessary to use a detector with a spatial resolution which is limited in an adapted way. However, this leads to an undesired reduction in the detective quantum efficiency (DQE) at high spatial frequencies.

The post-published German patent application DE 101 51 568 discloses a method for applying an antiscatter grid to an x-ray detector in the case of which a basic structure for the antiscatter grid is produced directly on the detector surface by means of a rapid prototyping technique such that absorbing regions of the antiscatter grid are situated in less sensitive intermediate regions of the x-ray detector. However, the risk exists in this method of damaging the x-ray detector when producing the antiscatter grid.

On the basis of this prior art, it is the object of the present invention to specify a method for producing and applying an antiscatter grid or collimator to an x-ray or gamma detector having matrixially arranged detector elements and with the aid of which it is possible to realize an arrangement of an antiscatter grid or collimator on an x-ray or gamma detector which permits image recording without moiré structures in conjunction with a high detective quantum efficiency.

The object is achieved by the method as claimed in patent claim 1. Patent claims 16 and 17 specify an x-ray and gamma detector having an antiscatter grid or collimator produced and applied in accordance with the method. Advantageous refinements of the method are the subject matter of the subclaims, or can be gathered from the following description and the exemplary embodiments.

In the present method, a basic structure for the antiscatter grid or collimator is firstly produced by means of a rapid prototyping technique, preferably with the aid of the technique of stereolithography, through which transmission channels and intermediate walls of the antiscatter grid or collimator are formed which have at least in a first direction a center-to-center spacing which is equal to or an integral multiple of the center-to-center spacing of the sensitive detection regions of the detector. The intermediate walls are subsequently coated with a material which strongly absorbs x-radiation and/or gamma radiation in order to finish the antiscatter grid or collimator. The basic structure coated in this way, that is to say the antiscatter grid or collimator, is subsequently applied to the detector surface, and connected to the latter, in such a way that at least the intermediate walls running perpendicular to the first direction, or their coating, are situated over the less sensitive intermediate regions of the detector surface. These intermediate regions of the detector surface which are less or not sensitive correspond to the regions in which the individual detector elements abut one another. Since the detector elements are additionally generally not sensitive to radiation over the entire surface, such insensitive intermediate regions arise in the edge regions of the individual detector elements.

The basic structure for the antiscatter grid or collimator is constructed in the case of the present method in such a way that at least in one direction the intermediate walls or the absorbing coating provided on the intermediate walls inside the transmission openings extend on one side of the intermediate walls over the intermediate regions on the detector. If, because of particularly narrow intermediate regions, the intermediate walls cannot be produced with a sufficiently small thickness, they can be situated in

the contact region with the detector surface at least partially over the intermediate regions. The intermediate walls extending in the other direction are preferably likewise arranged in such a way that they or
5 their coating are situated on one side over the non-sensitive intermediate regions of the detector surface. In both dimensions, the spacings of the intermediate walls can thereby assume the value of an integral multiple of the center-to-center spacing of the
10 sensitive detection regions of the detector. However, the center-to-center spacings of the intermediate walls preferably correspond in both mutually perpendicular directions to the corresponding center-to-center spacings of the sensitive detection regions.

15 By using a rapid prototyping technique when constructing the basic structure, very filigree structures can be produced with very high accuracy. In the rapid prototyping technique, 3D CAD designs, here
20 the geometry of the basic structure, are converted into volume data in the CAD system. The 3D volume model for the rapid prototyping is then divided into cross sections in a computer. The cross sections have a layer thickness of 100 μm or less. After the data have been
25 sent to a rapid prototyping system, the original shape is built up layer by layer. The present method in this case uses a rapid prototyping technique in which the layer construction is carried out by action of radiation, in particular laser radiation. Laser
30 radiation, specifically, offers the advantage of producing very filigree structures in this case.

In a preferred embodiment of the present method, the technique of stereolithography is used for constructing
35 the basic structure. In this method, a computer-controlled UV laser beam forms the respective contours of the individual layers of the 3D volume model of the basic structure on a liquid polymer resin. The resin is

cured under the action of the laser at the exposed points or areas. The component platform of the system is then lowered, and a new thin layer of photopolymer resin is applied. By repeating these steps, the complete geometry of the basic structure is successively constructed from the bottom upward. In one embodiment of the present method, it is also possible to use the technique of microstereolithography to produce the basic structure.

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By comparison with the post-published document specified in the introduction to the description, the production and application of an antiscatter grid or collimator directly onto the detector is simplified by the present method and can be realized more cost-effectively. Owing to the configuration and proposed application of the antiscatter grid or collimator in the specified way such that one or two of the total of four intermediate walls or their coating correspond sufficiently on one side with a boundary line between the detector pixels, the required dose when using this antiscatter grid or collimator is reduced. On the basis of this arrangement in regions which have a lesser sensitivity to the x-ray or gamma radiation, the intermediate walls or the absorbing coating provided thereon act to attenuate primary beams less or not at all. The proposed configuration and the application of the antiscatter grid or collimator principally in the said intermediate regions additionally excludes interference with the pixel structure of the detector.

The detectors virtually always have a filling factor which is smaller than 1. This holds, in particular, for a-Si detector surfaces coated with phosphor. Even in the case of detector surfaces coated with selenium, the filling factor differs from 1, particularly for small detection regions or pixels. Consequently, the quantum efficiency is primarily reduced in the regions between

the pixel surfaces. If the primary radiation is now attenuated only between the pixels by structures absorbing the scattered beams, this is more advantageous for achieving a high quantum efficiency than if these structures are arranged arbitrarily. Moiré interference between the pixels and the absorbing structure is thereby impossible. The present arrangement of the antiscatter grid or collimator thus permits the primary radiation to be more effectively rendered useful, since the unavoidable primary absorption of the antiscatter grid or collimator occurs chiefly in geometrical regions of the detector which make a reduced contribution to the image signal.

15 In a preferred embodiment of the present method, the end faces of the intermediate walls are kept free of the absorbing coating, or the absorbing coating possibly applied is removed from these end faces. Here, the end faces are understood to be the sides of the intermediate walls which face the detector, and the sides which are averted from the detector, that is to say the sides which are not situated inside the transmission channels. These end faces can be kept free, for example, by appropriate masks when the coating is being applied. However, the coating is preferably applied to the entire basic structure and subsequently removed by an appropriate chemical or mechanical method. If, moreover, use is made as material of the basic structure of a material which is substantially transparent to x-radiation and/or gamma radiation, the primary beam transmission of the antiscatter grid or collimator is substantially increased by this measure, since it is possible even in the material regions between the coated inner surfaces of the intermediate walls for appropriate primary radiation to pass through without attenuation or only slight attenuation and to contribute to the construction of the image. When use is made of the

technique of stereolithography for constructing the basic structure, such a refinement can be realized without difficulty by selecting a suitable polymer. Applying the absorbing layer can be performed in this case by various known methods, for example vapor deposition, by sputtering or by an electrolytic process. One possibility for applying the layer also consists in using sputtering to apply a thin metal layer which then serves as starting layer for subsequent electrolytic deposition of the layer.

In a further advantageous refinement of the present method, the antiscatter grid or collimator is applied to the detector surface and connected to the latter in such a way that in each case a corner region of the coating inside a transmission channel comes to be situated over a switching element of a detector element. Switching elements of this type, such as a diode or TFT, have no photosensitivity at all, and therefore make no contribution to the detection of radiation. Consequently, the attenuation of the primary radiation in this region has no great influence owing to the positioning of the corner regions of the coating over these switching elements.

Various techniques can be used for the adjusted application of the antiscatter grid or collimator to the detector surface. One technique consists in marking the precise desired position of the basic structure with reference to the pixels, situated therebelow, of the detector as fiducial markers or reference lines on the surface of the detector or on a protective layer which is applied to the scintillator. If appropriate, the desired position can also be projected on optically. The marking of the desired position can be controlled by means of infrared microscopy. Subsequently, the antiscatter grid or collimator is connected to the detector surface, for example by

bonding. In this case, the bonding is performed step by step such that a small subarea is firstly bonded after the position of the basic structure in relation to the pixels has been precisely set in this region.

5 Thereafter, the position in the adjacent subareas is corrected with the aid of visual support, for example under microscopic control and/or by using a projector which projects the desired position of the basic structure onto the mounting surface, and bonded further

10 subarea by subarea. For this procedure, the antiscatter grid or collimator can also be decomposed into smaller segments before being applied, for example by cutting with a laser beam. This facilitates the bonding, because then it is respectively necessary to adjust and

15 apply only small areas in each adhesion step. The cutting gaps produced by cutting the antiscatter grid or collimator must, however, be executed to be so small that they do not lead to any perturbing artifacts in the image.

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A further possibility for applying the antiscatter grid or collimator to the detector surface consists in using the flip-chip technique known from semiconductor technology. Instead of the chip, the antiscatter grid

25 or collimator is thereby applied to the detector surface instead of the substrate. Since the grid or segment to be bonded on is mechanically unstable, it is preferable to use a stable mechanical holder for the antiscatter grid or collimator or the segments. This

30 holder can consist, for example, of a partial height of the negative of the basic structure. The negative can be produced, for example, by saw slits in a metal plate. Instead of saw slits, it is also possible to design this holding plate with a plurality of adjusting

35 pins which can be inserted into the transmission openings of the basic structure.

A further possibility for positioning the antiscatter grid or collimator accurately in relation to the pixels of the detector consists in undertaking this adjustment in the presence of x-radiation or gamma radiation. Use
5 is made in this case of the fact that the position of the antiscatter grid or collimator in relation to the pixel arrangement is optimum whenever the output signal of the detector assumes the maximum in this position. For this purpose, the position of the antiscatter grid
10 or collimator is firstly adjusted, and bonded, in a relatively small segment of the detector. Thereafter, the further segments are successively adjusted and bonded.

15 Various adhesives can be used in the present invention. Thus, for example, an adhesive liquid can be sprayed in the region to be fixed onto the detector surface, or else the antiscatter grid or collimator, before these are interconnected. A further possibility consists in
20 using a soft, thermally melting adhesive which is applied before mounting to the detector surface and/or the antiscatter grid or collimator. By using a local heat source such as hot air, for example, the adhesive is only briefly melted with the antiscatter grid or
25 collimator correctly positioned, in order to interconnect the grid or collimator and detector.

Of course, any desired detector types are possible as x-ray or gamma detectors to which an antiscatter grid
30 or collimator can be applied using the present method. Thus, for example, it is possible to use detectors having a photoconducting material, or else detectors to which a scintillator layer has been applied.

35 The present method will be explained again briefly below with the aid of exemplary embodiments in conjunction with the drawings, without restricting the general inventive concept.

- Figure 1 shows the situation when recording x-ray images of an object using an antiscatter grid;
- 5 Figure 2 shows the situation when recording the object in nuclear medicine using a collimator;
- Figure 3 shows a representation of the technique of stereolithography when constructing a structure;
- 10 Figure 4 shows an example of carrying out the present method;
- Figure 5 shows an example of the arrangement, realized using the method, of antiscatter grid or collimator and detector;
- 15 Figure 6 shows a second example of the arrangement, realized using the method, of antiscatter grid or collimator and detector; and
- 20 Figure 7 shows a system for applying the antiscatter grid or collimator.

The typical situation when recording an x-ray image of an object 3 in x-ray diagnosis is represented schematically with the aid of Figure 1. The object 3 lies between the tube focus 1 of an x-ray tube, which may be regarded as an approximately point x-ray source, and a detector surface 7. The x-rays 2 emitted from the focus 1 of the x-ray source propagate in a straight line in the direction of the x-ray detector 7, and in doing so pass through the object 3. The primary beams 2a striking the detector surface 7, which pass through the object 3 on a straight line starting from the x-ray focus 1, cause, on the detector surface 7, a positionally resolved attenuation value distribution for the object 3. Some of the x-ray beams 2 emitted from the x-ray focus 1 are scattered in the object 3. The scattered beams 2b created in this case do not

contribute to the desired image information and, when they strike the detector 7, they significantly impair the signal-to-noise ratio. In order to improve the image quality, an antiscatter grid 4 is therefore
5 arranged in front of the detector 7. This antiscatter grid 4 has transmission channels 5 and a basic structure 6 which consists of an absorbing material in the case of the prior art represented. The transmission channels 5 are aligned in the direction of the tube
10 focus 1, so that they allow the incident primary radiation 2a on a straight-line path to strike the detector surface. Beams not incident in this direction, in particular the scattered beams 2b, are blocked or significantly attenuated by the absorbing intermediate
15 walls of the basic structure 6. However, on the basis of the previously known production techniques, the absorbing intermediate walls can be produced only with a particular minimum thickness, so that a significant part of the primary radiation 2a is therefore also
20 absorbed and does not contribute to the image result.

Figure 2 shows the situation when recording images in nuclear diagnosis. The body 3 to be examined, in which an organ 3a is indicated, can be seen in the figure. By
25 injection of a medium which emits gamma radiation, and which concentrates in the organ 3a, gamma quanta 8a are emitted from this region and strike the detector 7, an Anger camera. By means of the collimator 4 arranged in front of the detector 7, which has transmission
30 channels 5 aligned in a straight line between intermediate walls of the basic structure 6 that absorb gamma radiation, the projection direction of the respective image recording is set. Gamma quanta 8b which are emitted in other directions or are scattered,
35 and which do not arrive on a straight-line path from this projection direction, are absorbed by the collimator 4. In this technique as well, however, a significant part of the primary radiation 8a is still

absorbed because the absorbing intermediate walls of the basic structure 6 are not arbitrarily thin.

The present invention may be implemented by using a method which permits very precise manufacture of antiscatter grids or collimators with very thin intermediate or separating walls between the transmission channels 5. In this case, a rapid prototyping technique is used for producing the antiscatter grid or collimator. An example of such a technique is stereolithography, as illustrated with the aid of the representation in figure 3. In this technique, a UV laser beam 12 is directed onto the surface of a liquid UV-crosslinkable polymer 10, which is situated in a container 9. The UV laser beam 12 is moved, with the aid of a three-dimensional volume model of the basic structure 6 to be made, over the surface of the liquid polymer 10 in order to construct the basic structure 6 layer-wise. After a layer has been solidified, it is lowered by a further layer thickness using a construction platform 11, so that the UV laser 12 can solidify the next layer according to the three-dimensional volume model. In this way, the basic structure 6 is constructed layer by layer from the crosslinked UV-cured polymer 10. Owing to the good focusability of the UV laser beam 12, very filigree structures can in this case be produced with very high accuracy. The basic structure 6 may be constructed directly on the construction platform 11 or on an additional support plate (not represented in the figure). Furthermore, the technique of stereolithography may also be used to directly construct a baseplate, on which the basic structure 6 is then formed according to the desired geometry.

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Figure 4 shows an example of carrying out the present method. In this embodiment, a basic structure 6 is produced accordingly for the antiscatter grid or

collimator to be made from a structural material which is substantially transmissive to x-radiation, for example a UV-cured polymer. The basic structure 6 is subsequently coated (coating step 13; figure 4a) with an absorbing material of high atomic number. Respectively to be seen in the individual part-figures 4a - 4b are a detail of the basic structure 6 with the transmission channels 5 and the intermediate walls 6a and of the antiscatter grid or collimator, respectively, in cross section in the upper part, and in plan view in the lower part. This coating step 13 results in a basic structure 6 with a strongly absorbing coating 14 which is applied both to the inner surfaces of the intermediate walls 6a in the transmission channels 5, and to the top side and underside 6b of the basic structure 6, that is to say to the end faces of the intermediate walls 6a (figure 4b).

Finally, the coating 14 is removed at the end faces of the intermediate walls 6a by a chemical or mechanical method step, for example a chemical etching step 15 (figure 4c). An antiscatter grid or collimator is obtained in this way which has transmission channels 5 that are situated between the absorbing coatings 14 of the intermediate walls 6b of the basic structure 6 (figure 4d). The intermediate walls 6b likewise form with the coatings 14 delimiting them transmission channels for the primary radiation which have a very high aspect ratio.

For application of the coating 14, a thin copper layer may, for example, firstly be applied to the surface of the basic structure 6 by sputtering. The thickness of this copper layer lies in the 1/10 μm range. A lead-tin alloy is finally applied to this layer by electrolytic deposition. The thickness of this alloy is designed so

as to achieve the necessary absorption of the secondary radiation.

After finishing the antiscatter grid or collimator, the latter is bonded onto the detector 7 in such a way that the intermediate walls 6a are situated over less sensitive intermediate regions 16b of the detector (figure 4e). A detail of the detector 7 with the matricially arranged detector elements 16 is illustrated in figure 4e. The detector elements 16 form a detector surface having regions 16a which are sensitive to x-rays or gamma rays, and insensitive intermediate regions 16b.

Figure 5 shows a result of the method illustrated in figure 4. To be seen in plan view in the figure is a detail of an x-ray detector 7 with the matricially arranged detector elements 16 and the antiscatter grid bonded thereon. The detector elements 16 are assembled from the radiation-sensitive detector regions 16a and radiation-insensitive intermediate regions 16b. The detector elements 16 are also referred to as pixels. Such an x-ray detector 7 can have, for example, 3000 x 3000 pixels with dimensions of 143 x 143 μm . To be seen in the left-hand corner region of each detector surface is a switching element 17, for example a TFT, which does not contribute to the detection of radiation. Figure 5 shows an exemplary embodiment in which a wall thickness of approximately 20 μm was implemented for the intermediate walls 6a of the basic structure 6 of the antiscatter grid. As may be seen from the figure, the antiscatter grid is positioned with the intermediate walls 6a over the radiation-insensitive regions 16b of the detector 7. The absorbing coating 14 is present here only inside the transmission channels 5, and so attenuation of the primary radiation occurs only at this point. The wider wall regions 6a between this coating 14 have a

relatively high transmission for the primary radiation, and so these can likewise be regarded as transmission regions. These intermediate walls 6a are formed in the present example from a polymer which attenuates the x-radiation only insubstantially. Arranged at the crossing point of the radiation-insensitive regions 16b of the detector 7 are the corresponding switching elements 17 over which corner regions of the absorbing coating 14 are situated in the present example.

10 The production of such an arrangement of antiscatter grid or collimator and detector can be implemented simply and cost-effectively. In particular, it is advantageous in this case that a fault in the
15 production of the grid does not lead automatically to the destruction of the entire detector. By keeping the end faces of the intermediate walls 6a free of the absorbing coating 14, it is possible to achieve a better primary beam transparency than with conventional
20 grids. The permanent connection of the grid and the detector in conjunction with the partial correspondence of the structures in the intermediate walls 6a and pixel boundaries achieves an improved sensitivity of the detector, and so dosage can be saved. Moiré
25 interference is avoided in the same way by this configuration. The transmission channels 5 illustrated in white in the figure are generally filled with air or gas.

30 Figure 6 shows a further example, in the case of which the intermediate walls 6a of the antiscatter grid have a thickness of approximately 70 μm , and can therefore be produced very easily by stereolithography. In this case, as well, a similar primary beam transparency is
35 achieved as in the case of the configuration of figure 5, since the absorber coating 14 on the end faces of the intermediate walls 6a is removed, and thus

primary radiation can also strike the detector 7 at these points.

Figure 7 shows a system for applying the finished antiscatter grid or collimator to the detector by means of flip-chip technology. The exact position of the antiscatter grid or collimator picked up with the aid of the manipulator arm 18, or else of a segment thereof, is controlled via a microscope 19 with the aid of which two images, superimposed by means of a beam splitter 20, of the detector surface and of the antiscatter grid or collimator can be viewed and can be brought into coincidence by displacement. If necessary, it is also possible to carry out a limited rotation of the detector for this purpose. If the reciprocal position is adjusted, the antiscatter grid is applied to the prepared bonded joint in an accurately positioned fashion by means of the manipulator. An appropriately designed carrier plate with corresponding depressions or holding pins is preferably used for holding the grid.

Patent Claims

1. A method for producing and applying an antiscatter grid or collimator to an x-ray or gamma detector having
5 matricially arranged detector elements (16) which form a detector surface with detection regions (16a) sensitive to x-radiation and/or gamma radiation and less sensitive intermediate regions (16b), in the case of which method a basic structure (6) is firstly
10 produced for the antiscatter grid or collimator by means of a rapid prototyping technique through which transmission channels (5) and intermediate walls (6a) of the antiscatter grid or collimator are formed which have at least in a first direction a center-to-center
15 spacing which is equal to or an integral multiple of a center-to-center spacing of the sensitive detection regions (16a) of the detector, the intermediate walls (6a) are subsequently coated with a material (14) which strongly absorbs x-radiation and/or gamma radiation in
20 order to finish the antiscatter grid or collimator, and finally the antiscatter grid or collimator is applied to the detector surface and connected to the detector surface in such a way that at least the intermediate walls (6a) running perpendicular to the first
25 direction, or their coating (14), are situated over the less sensitive intermediate regions (16b) of the detector surface.

2. The method as claimed in claim 1, characterized in
30 that the method of stereolithography is used as rapid prototyping technique.

3. The method as claimed in claim 1 or 2, characterized in that the geometry of the basic
35 structure (6) is selected corresponding to the matricial arrangement of the detector elements (16) so as to produce a cellular antiscatter grid or collimator in the case of which the arrangement of transmission

channels (5) transmissive to x-radiation and/or gamma radiation corresponds to the arrangement of the sensitive detection regions (16a).

5 4. The method as claimed in one of claims 1 to 3, characterized in that the basic structure (6) is produced from a material which is substantially transparent to x-radiation and/or gamma radiation, and end faces of the intermediate walls (6a) are kept free
10 of the coating with the absorbing material (14).

5. The method as claimed in one of claims 1 to 3, characterized in that the basic structure (6) is produced from a material which is substantially
15 transparent to x-radiation and/or gamma radiation, and the coating with the absorbing material (14) is removed from end faces of the intermediate walls (6a).

6. The method as claimed in one of claims 1 to 5,
20 characterized in that the coating is performed by sputtering and/or electrolytic deposition.

7. The method as claimed in one of claims 1 to 6, characterized in that the antiscatter grid or
25 collimator is applied to the detector surface and connected to the detector surface in such a way that in each case a corner region of the coating with the absorbing material (14) of a transmission channel (5) is situated over a switching element (17) of a detector
30 element (16).

8. The method as claimed in one of claims 1 to 7, characterized in that the antiscatter grid or collimator is bonded to the detector surface.

35 9. The method as claimed in claim 8, characterized in that the bonding is carried out sequentially for juxtaposed subareas of the detector surface.

10. The method as claimed in claim 9, characterized in that before being applied the antiscatter grid or collimator is divided into segments which correspond to the subareas and are applied sequentially in an individual fashion.

11. The method as claimed in one of claims 1 to 8, characterized in that the antiscatter grid or collimator is applied to the detector surface with the aid of a flip-chip technique.

12. The method as claimed in claim 11, characterized in that the antiscatter grid or collimator is applied to the detector surface with the aid of a holder supporting the basic structure (6).

13. The method as claimed in one of claims 1 to 12, characterized in that for bonding purposes an adhesive liquid is sprayed onto the detector surface and/or the antiscatter grid or collimator.

14. The method as claimed in one of claims 1 to 12, characterized in that for bonding purposes a thermally melting adhesive is applied to the detector surface and/or the antiscatter grid or collimator and is briefly melted by a heat source after the application of the antiscatter grid or collimator to the detector surface.

15. The method as claimed in one of claims 1 to 14, characterized in that the basic structure (6) is constructed so as to produce a focused antiscatter grid or collimator.

16. An x-ray detector having matricially arranged detector elements and an antiscatter grid with transmission channels (5) and intermediate walls (6a)

with an absorbing coating (14), which is produced and applied as claimed in one of patent claims 1 to 15.

17. A gamma detector having matricially arranged
5 detector elements and a collimator with transmission channels (5) and intermediate walls (6a) with an absorbing coating (14), which is produced and applied as claimed in one of patent claims 1 to 15.

Abstract

Method for producing and applying an antiscatter grid or collimator to an x-ray or gamma detector

The present invention relates to a method for producing and applying an antiscatter grid or collimator to an x-ray or gamma detector having matricially arranged detector elements (16) which form a detector surface with detection regions (16a) sensitive to x-radiation and/or gamma radiation and less sensitive intermediate regions (16b). In the method, a basic structure (6) is firstly produced for the antiscatter grid or collimator by means of a rapid prototyping technique through which transmission channels (5) and intermediate walls (6a) of the antiscatter grid or collimator are formed which have at least in a first direction a center-to-center spacing which is equal to or an integral multiple of a center-to-center spacing of the sensitive detection regions (16a) of the detector (7). The intermediate walls (6a) are coated with a material (14) which strongly absorbs x-radiation and/or gamma radiation in order to finish the antiscatter grid or collimator. Subsequently, the antiscatter grid or collimator is applied to the detector surface and connected to the detector surface in such a way that at least the intermediate walls (6a) running perpendicular to the first direction, or their coating (14), are situated over the less sensitive intermediate regions (16b) of the detector surface. A detector having an antiscatter grid or collimator in which no moiré interference occurs can be realized in a simple way using the present method.

Figure 6

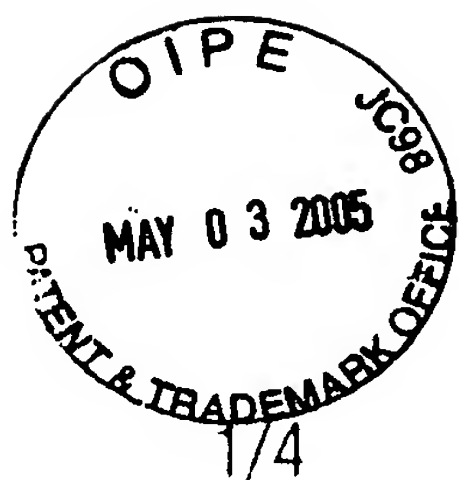


FIG 1

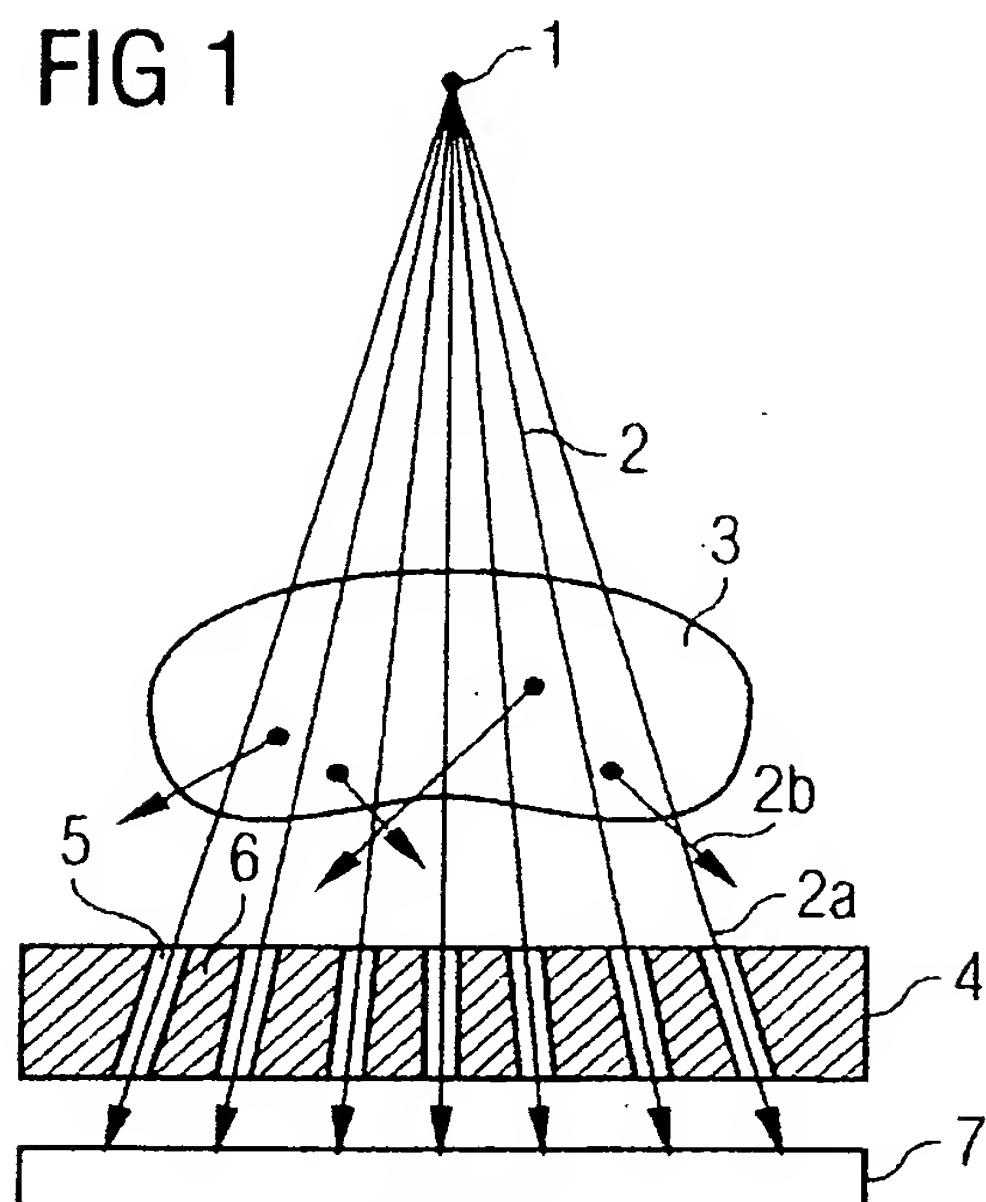


FIG 2

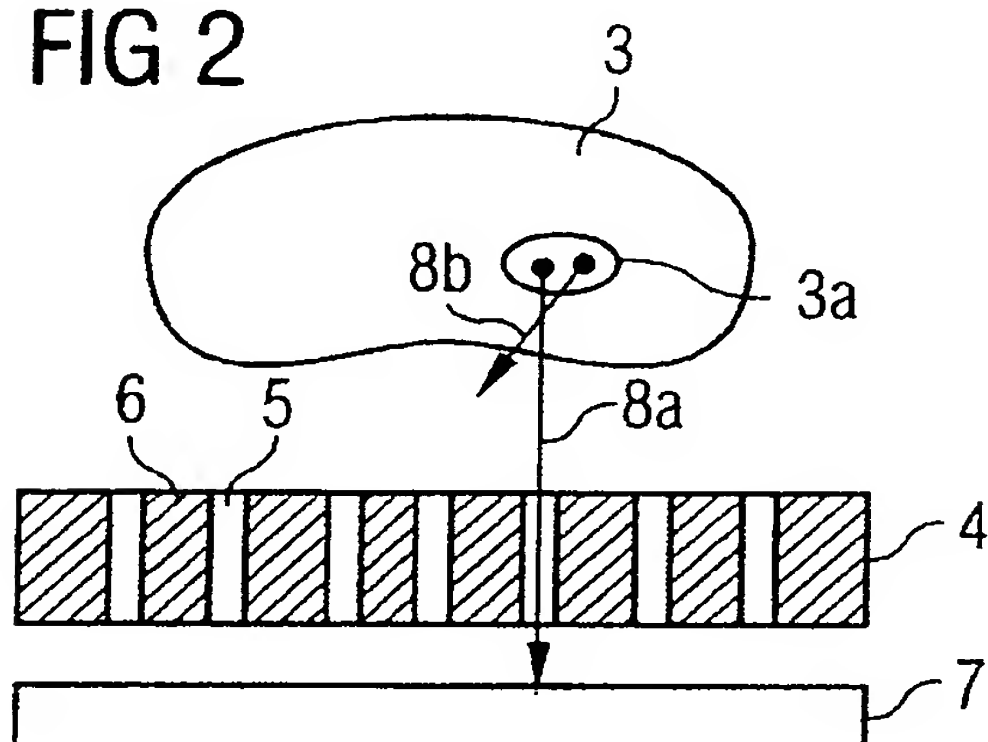


FIG 3

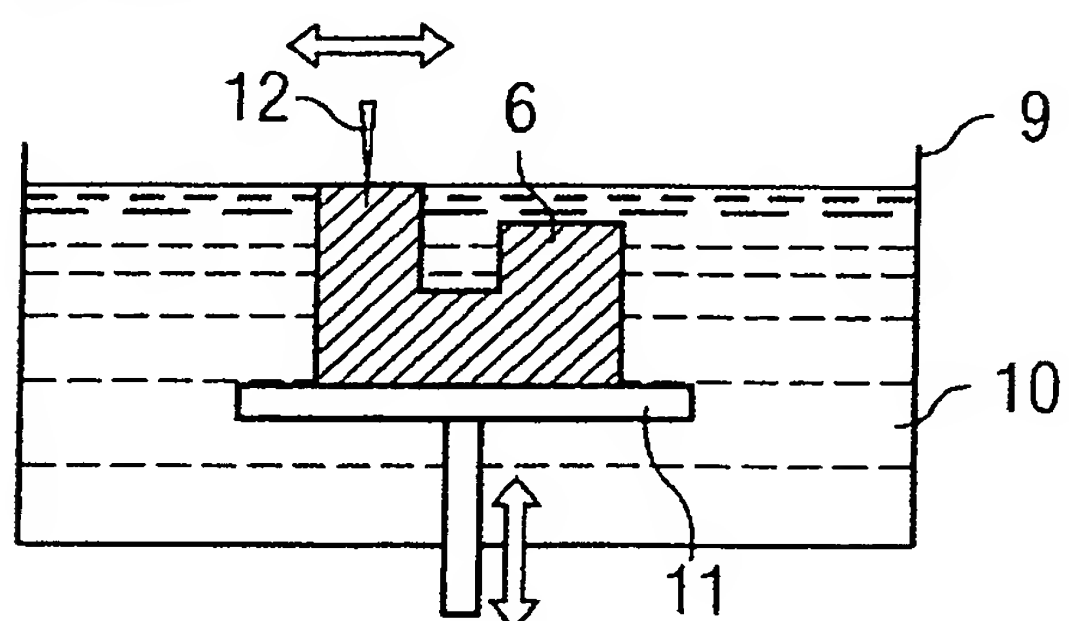


FIG 4A

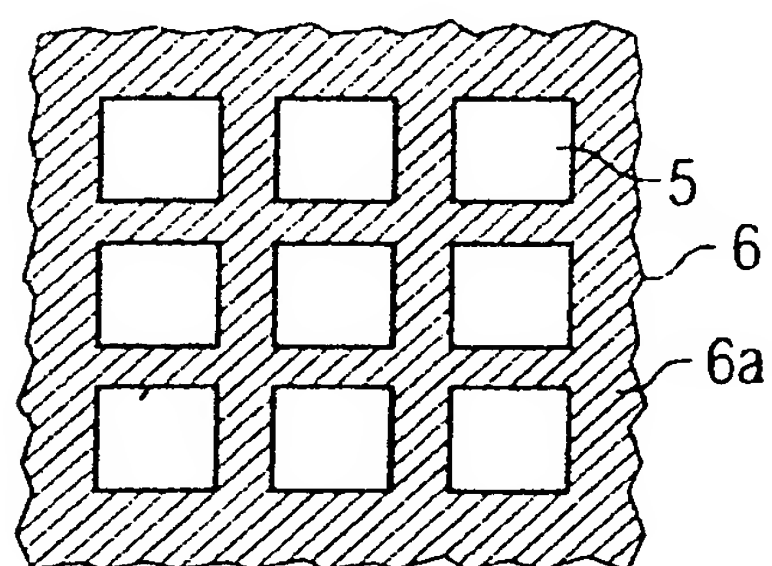
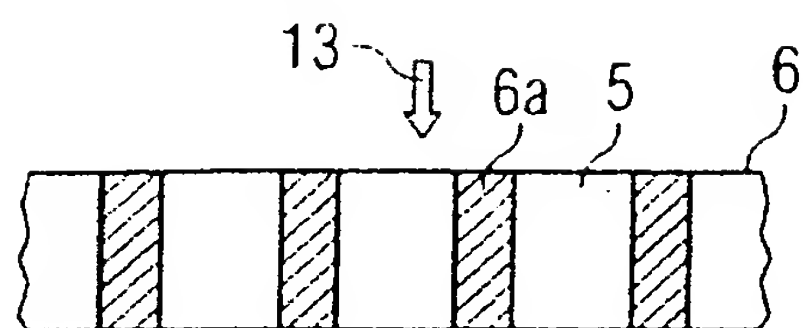


FIG 4B

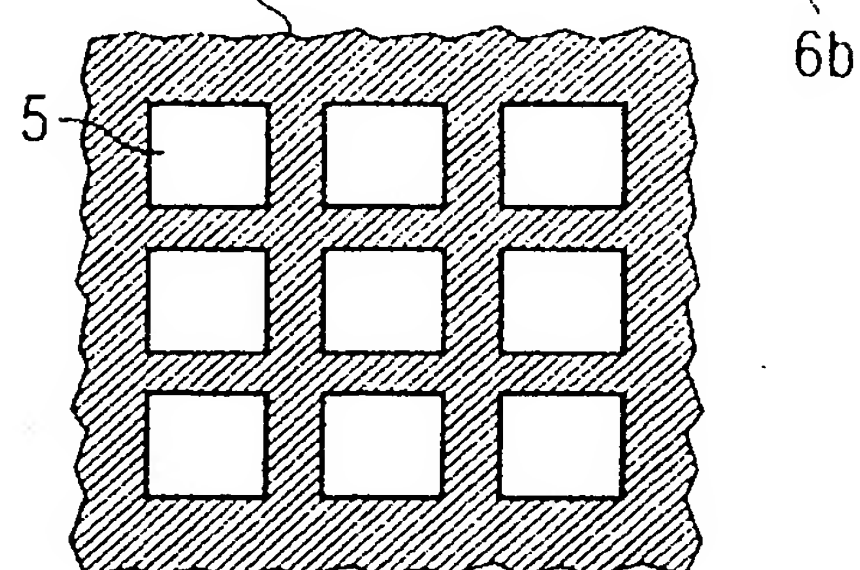
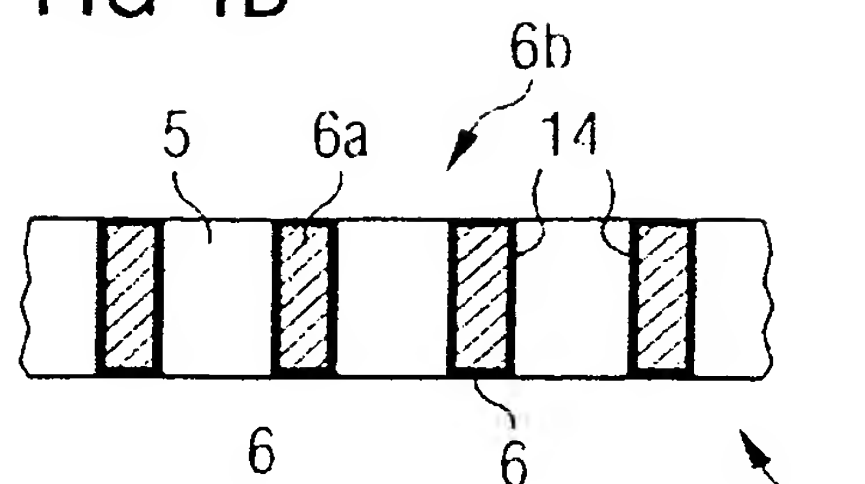


FIG 4C

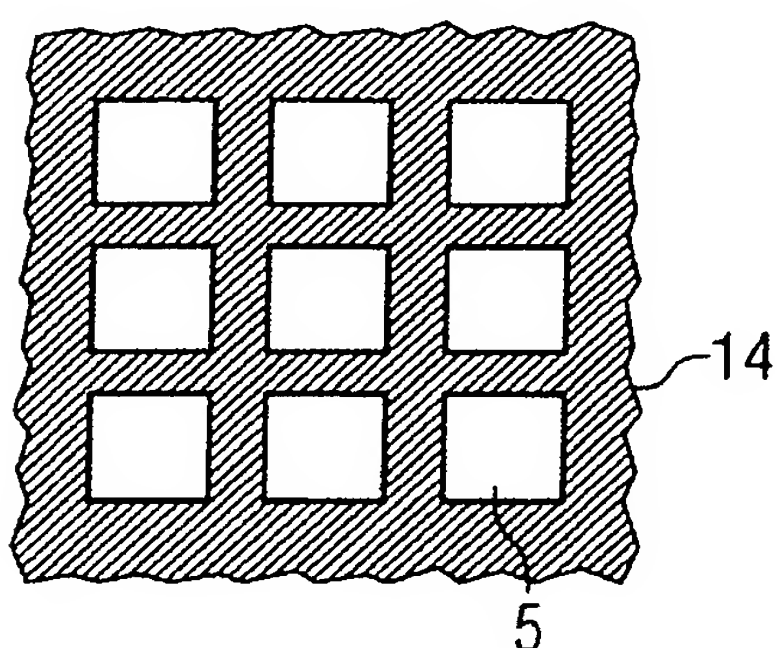
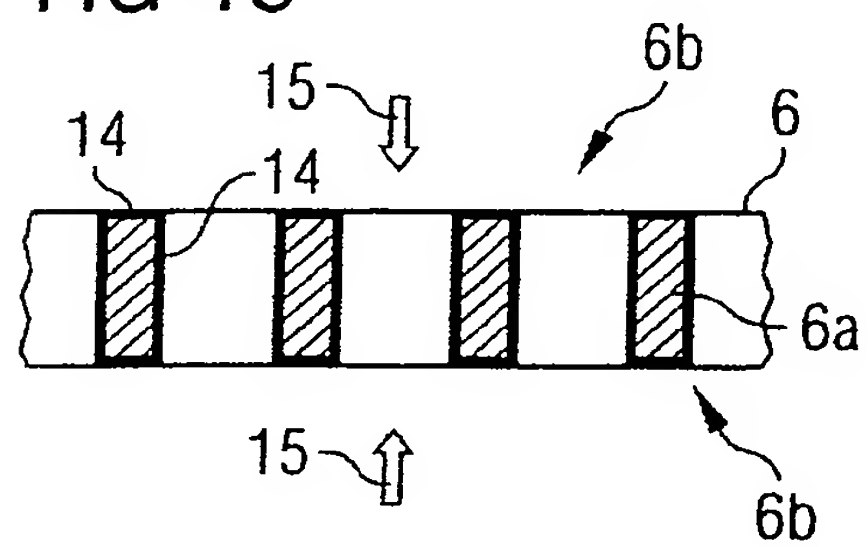


FIG 4D

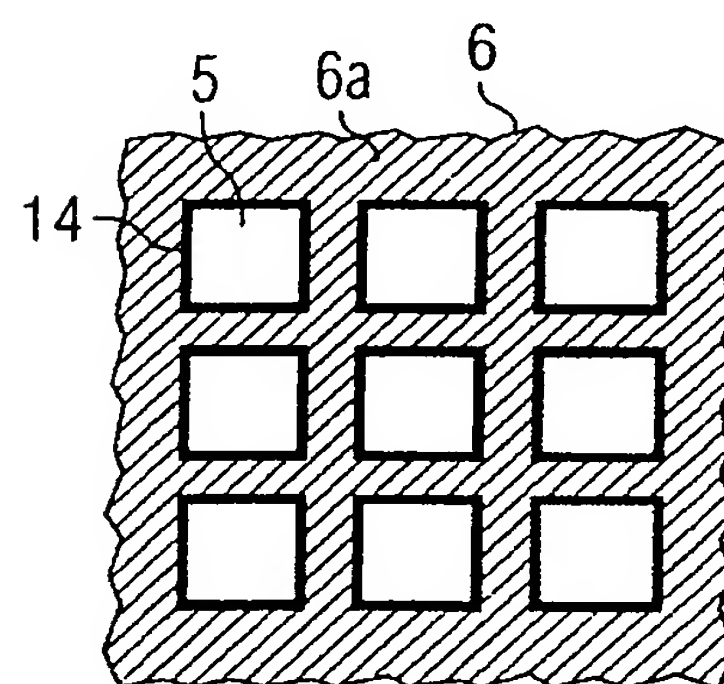
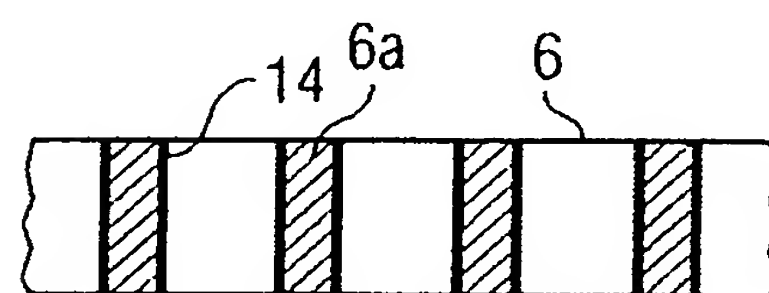


FIG 4E

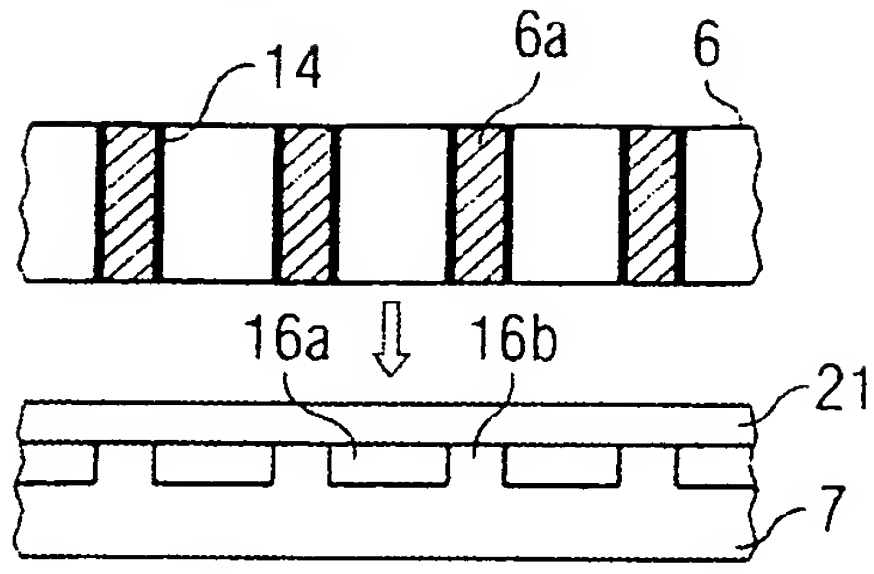


FIG 5

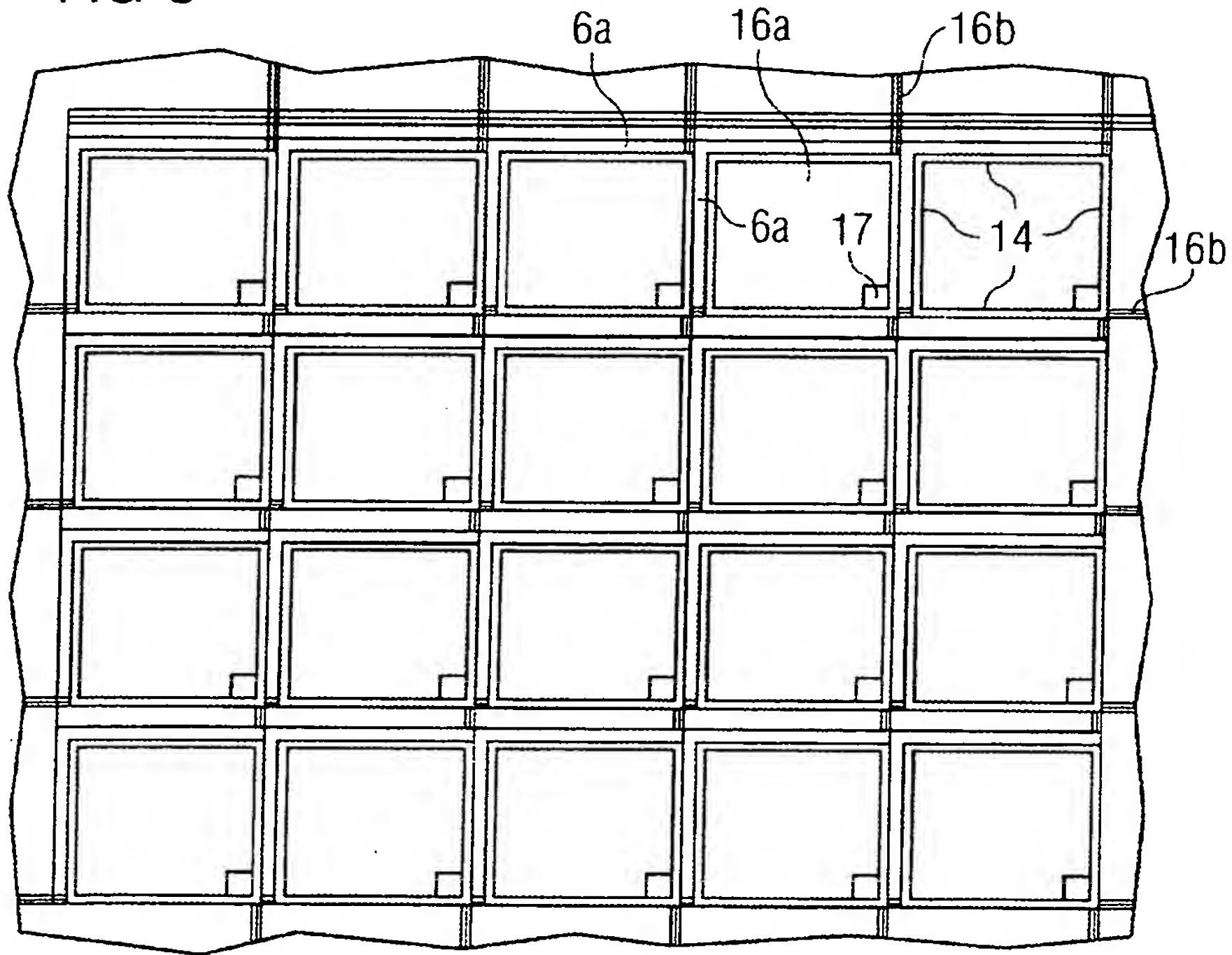


FIG 6

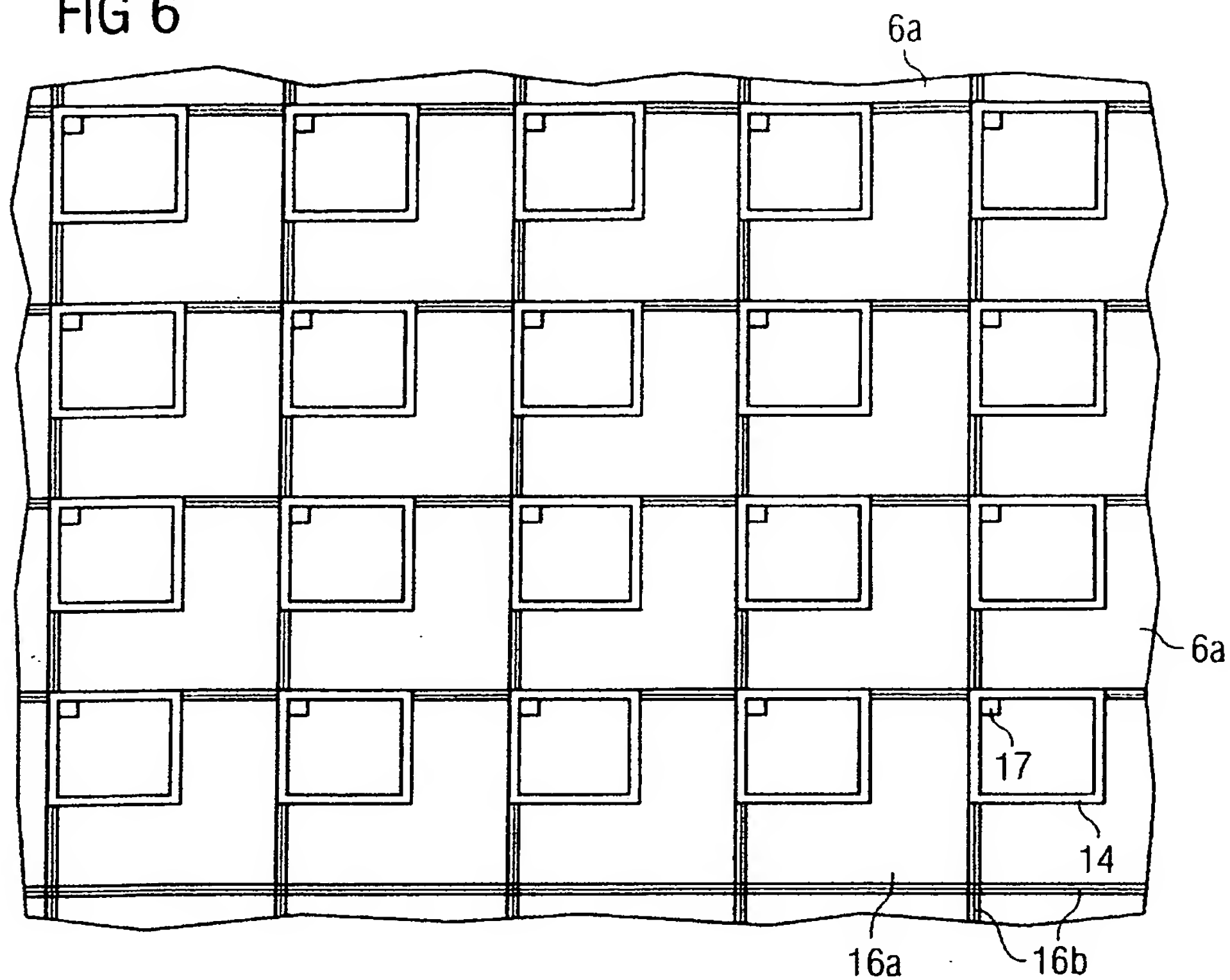


FIG 7

